

Passive elastic contribution of hip extensors to joint moments during walking in people with low back pain

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Abstract

Background. It has been found that alterations in passive muscle properties may be associated with low back pain, and these may be responsible for the altered gait parameters often observed in subjects with back pain. The purpose of the present study

was to assess total hip and passive hip extensor moments in people with or without low back pain during the hip flexion component of walking.

Methods. 52 subjects volunteered for this study (low back pain group, $n = 25$ (male $n = 13$, female $n = 12$), control group, $n = 27$ (male $n = 15$, female $n = 12$)). Passive hip moments were calculated using an adapted force transducer during supine testing. A biomechanical model and predictive equation were used to calculate passive hip moments during walking. Total hip moments were calculated with the use of a 9 camera, 3-D motion-capture system.

Findings. Independent samples t-tests demonstrated no significant differences between groups for gait parameters or hip or knee angles. Results of the ANOVAs demonstrated significant differences in passive hip flexor moments during the second half of hip flexion ($P < 0.05$). There were also significant differences in hip power and work done during peaks of power absorption and the second peak of power generation ($P < 0.05$).

Interpretation. The present data demonstrates that subjects with low back pain have altered passive hip extensor and total power and work done during walking compared with healthy controls. Biomechanical models should include individual measurements of passive joint moments.

Passive elastic contribution of hip extensors to joint biomechanics during walking in people with low back pain

1
2 Low back pain (LBP) is one of the leading causes of disability globally [1]. Clinical
3 assessments of LBP patients often include tests of hip extensor extensibility [2]. Tests
4 can include assessment of knee extension angle and sacral angle, the sit and reach test
5 [3], and the straight leg raise test [3, 4]. However, the relevance of hip muscle extensibility
6 to LBP and any relationship to movement remains unclear [5]. There is a growing interest
7 in including more comprehensive assessments of joint and muscle resistive properties,
8 due to the lack of consensus with assessing extensibility alone [6, 7].

9 Joint passive resistance is a property of the non-contractile tissues, such as the
10 tendon, sarcolemma, endomysium, perimysium and epimysium [6, 8-10], structural
11 proteins such as titin [11], and inactive muscle fibres. Active resistance is a property of
12 contracting muscle fibres. Some investigators report musculotendinous extensibility
13 being moderately related to passive stiffness, and weakly related to active stiffness [9].
14 An investigation by Halbertsma *et al.*[7], reported an association between hamstring
15 extensibility and LBP, but no differences in passive stiffness between LBP patients and
16 controls. Overall, there is a lack of agreement in the literature as to whether passive
17 muscle resistance is related to extensibility [9, 12]. Further, any relationship between
18 extensibility, passive and active stiffness and activities of daily living (ADLs) in LBP
19 patients remains to be determined.

20 It has been reported that muscle resistive properties may be altered in subjects
21 with LBP [6, 13, 14], and these could be responsible for the reduced leg swing, step length
22 and gait velocity often observed in LBP subjects when compared with healthy controls
23 [15-18]. During the late swing phase of walking, the activation of biceps femoris is

increased in LBP [18], indicating altered active factors. Identifying alterations to passive or active components may be a useful tool for the clinical assessment of LBP, and the first stage in developing effective physical therapy-based treatment strategies.

The purpose of the present study was to assess total hip and passive hip extensor moments in people with LBP during the hip flexion component of walking, and to compare them with pain-free controls. Passive hip moments were calculated as a product of hip and knee angle using a dynamic biomechanical model. Further comparisons were made of total hip power and work done during hip flexion and the complete gait cycle.

2 Methods

2.1 Participants

Fifty-two subjects volunteered for this study. Subjects were excluded if they were pregnant or had any tumours, rheumatological or musculoskeletal disorders, tuberculosis, or an injury or infection of the spine, hips or knees during the 3 months prior to their participation. Subjects were also excluded if they had a history of any dislocation or surgery of the spine or lower limbs. Female subjects were only eligible for testing during the 7 days following the first day of menstruation, to control for any potential effects of the ovarian cycle.

Subjects meeting the inclusion criteria were divided into groups according to if they had suffered with chronic, non-specific low-back pain (LBP group, $n = 25$ (male $n = 13$, age = 34 (SD 8.53) years female $n = 12$ age = 30 (SD 7.96) years) for at least 6 weeks,

including at least one episode during the week of the study, or were back-pain-free (NBP group, $n = 27$ (male $n = 15$, age=29 (SD 7.78) years, female $n = 12$, age=33 (SD 8.78) years). Subjects in the NBP group needed to have been without back pain during the 6 months prior to the study.

Following consent to participate, subjects were required to complete a medical screening form and International Physical Activity Questionnaire (short form) (IPAQ-SF). LBP subjects were required to complete a Roland Morris Disability Questionnaire (RMDQ) and to rate their level of pain on a visual analogue scale (VAS). The study was approved by the ethics committees of both the University of Roehampton and the British College of Osteopathic Medicine.

2.2 Experimental Setup

Passive hip extensor moments were calculated during leg raising tests using an adapted force transducer, comprising a bi-axial cantilever load cell (QLA263, Futek, Irvine, US), and 2 analogue electro-inclinometers (PTAM27, ASM, Moosinning, Germany). The force transducer was inserted into a custom-built ankle brace designed to house the transducer with minimal friction, whilst maintaining the ankle in neutral. Four knee braces were pre-formed to secure the knee at 180, 170, 160 and 140 degrees, where 180 degrees refers to the knee in full extension. An additional two electro-inclinometers were secured using straps to the thigh and shank, to measure hip angle and angular acceleration, and knee angle, respectively. Two single differential surface electromyography (EMG) electrodes (SX230, Biometrics, Newport, UK), were placed over the biceps femoris and rectus

femoris, in accordance with the SENIAM guidelines for electrode placement. The EMG signals were used for real-time feedback to ensure no activity, and not for subsequent analysis.

The analogue signals from the load cell were pre-amplified (CSG110, Futek, Irvine, US) with 15 VDC for each input and analogue-digital converted. All analogue signals from the load cell and inclinometers were acquired at 50 Hz, and from the EMG electrodes at 1000 Hz, using a data acquisition unit (Datalink, DLK900, Biometrics, Newport, UK). Load cell and inclinometer data was digitally filtered at 2 Hz using a low-pass Butterworth filter, and saved to a personal laptop computer (Dell Precision, M4500, Dell, Bracknell, UK) for processing with Matlab programming software (Version 7.3, Mathworks, Natick, US).

Subjects were required to lie supine on a massage table. In accordance with the procedures of Lee and Munn [19], the test leg was passively raised 10 times to precondition the tissues, and to account for variability in activity levels between subjects immediately prior to testing (the test set-up is shown in figure 1). Supine passive leg raises were performed 3 times with each of the 4 knee braces, with 1 minute rest between tests with the same knee brace, and 2 minutes between different braces. During testing the subject was required to verbally indicate if and when they felt an onset of stretch-related pain. Each test would cease upon the onset of pain or deviations in EMG muscle activity above resting baseline level.

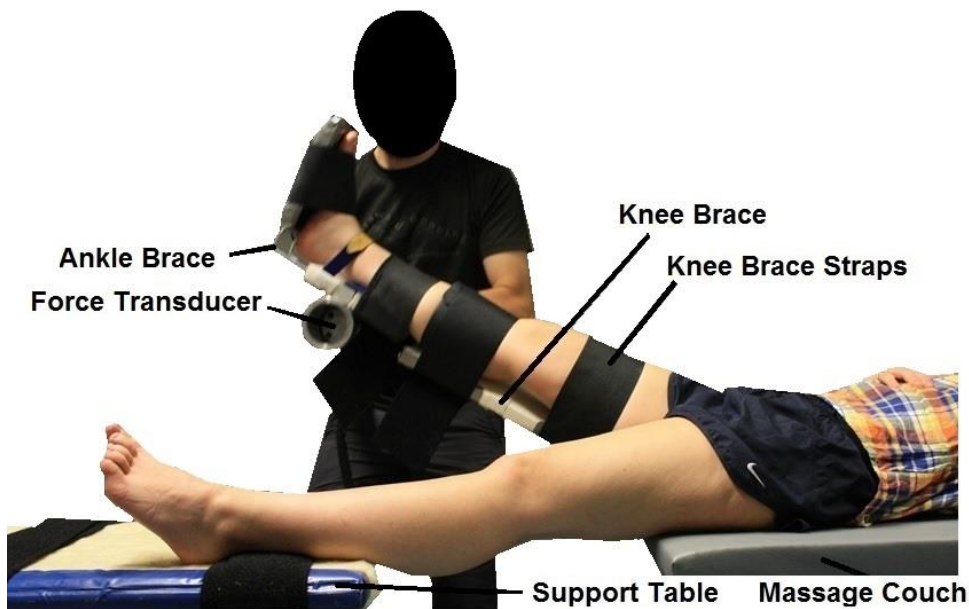


Figure 1. Subject test set-up for passive hip moment assessment, showing placement of force transducer, ankle brace and one of the 4 knee braces used to maintain the knee at a predetermined angle (180 degrees in the example above).

Total hip moments were calculated during level walking with the use of a 9 camera, 3-D motion-capture system (T-series, Vicon, Oxford, UK) and 2 force plates (9281CA, Kistler, Winterthur, Switzerland). 35 retro-reflective markers were placed on each subject in accordance with the placements used by previous researchers [20-22]. Motion capture data was sampled at 100-Hz and force plate data at 1000-Hz. Data was initially assessed via the Nexus software programme (Vicon Nexus version 1.8, Vicon, Oxford, UK), before being transferred to Microsoft Excel (2010, Microsoft, Redmond, US) and Matlab for further processing.

Subjects were required to walk along a 10-metre walkway in view of the motion capture cameras and over the force plates at their normal walking speed. The tester observed foot contact onto the force plates and ensured a minimum of 5 walks contained

both left and right clear foot strikes. Walking was at each subject's normal pace without reasonable likelihood of fatigue during testing.

2.3 Data Processing

A data processing pipeline was created in Vicon Nexus to perform standard data modelling of the walking trials. The pipeline included Woltring filtering and gap filling. Following determination of marker trajectories data was smoothed using a low-pass Butterworth filter at 6 Hz. Total moment-angle data for the hip was established for each complete gait cycle, and data was used from the first 5 complete tests on each leg. The coefficient of multiple correlation (CMC) of the moment-angle curves were calculated. The coefficient of variation (CV) was also calculated to assess intra-subject gait cycle variability.

Passive hip extensor moments were calculated based upon the dynamic biomechanical model developed by Lee and Munn [19]:

$$M = F_x y_f + F_y x_f + m_{leg} \ddot{x}_{cg} y_{cg} - (m_{leg} g + m_{leg} \ddot{y}_{cg}) x_{cg} - m_{leg} k^2 \ddot{\theta}$$

where F_x , F_y , are the forces applied to the leg to flex the hip joint, x_f , y_f , are the locations of force application to the leg, m_{leg} is the mass of the leg, g , is acceleration due to gravity, x_{cg} , y_{cg} , refer to the location of the centre of mass of the leg and k is the radius of gyration. \ddot{x}_{cg} , \ddot{y}_{cg} refer to the acceleration of the leg centre of mass, and $\ddot{\theta}$ is the angular

1 acceleration of the leg. Dempster [23] body segment parameter ratios were used for both
2 the total and passive moment calculations.

3 Passive hip moments are a property of the length of moment arm and the
4 amount of soft tissue, and so data was normalised to body mass and height. Hip moment-
5 angle data was originally determined from 0 degrees (hip neutral) to maximum hip range
6 of motion. Because the number of samples was reduced as hip angle increased, the
7 range 0 to 60 degrees was used for analysis as a majority of subjects achieved this. The
8 mean curve from 3 leg raises, smoothed using cubic spline interpolation and fitted with
9 an exponential function, was used for further analysis.

10 To establish a predictive equation for passive hip extensor moments based upon
11 a combination of hip and knee angles, 3-D surface plots were generated for assessing
12 the changes in hip moments with hip angle, and at different knee angles. In agreement
13 with other researchers [19], passive hip moments were found to increase exponentially
14 as a function of hip angle. By introducing a variable knee angle component, it was found
15 that this contributed a linear component to the equation. A surface fitting programme was
16 written for Matlab, and visual and residual analyses used to determine the most
17 appropriate equation coefficients for the individual datasets. From the present
18 investigation, the following equation was developed to predict passive hip moments
19 ($M_{passive}$) during flexion:

$$M_{passive} = a * \theta_{knee} + b * \exp^{c * \theta_{hip}} + d$$

1 where θ_{knee} is the angle at the knee, θ_{hip} the angle at the hip, and a , b , c , and d are the
2 equation coefficients.

3 Passive hip moments and 95% confidence intervals were recalculated from hip
4 neutral to maximum hip flexion angle, using the derived predictive equation. Root mean
5 squared error (RMSE) and adjusted r-squared were calculated to assess goodness of fit
6 of each surface-plot.

7 Hip moments were calculated using the conventional gait model, based upon the
8 Newington-Helen Hayes model. Total hip power was calculated from hip angular velocity
9 and total hip moments. All data was subsequently normalised to body mass and height.
10 Total hip moments were calculated at the peak hip flexor moment (FL), and the two hip
11 extensor peaks (Ext1, Ext2). Total hip power was calculated at the two peaks of power
12 generation (H1, H3) and peak of power absorption (H2). Mechanical work done at H1
13 and H3 was calculated by integrating the complete positive portions of the corresponding
14 power generation curves. Work done was also calculated for the full negative power
15 absorption curve (H2). Passive hip extensor moments and total hip moments, power and
16 work done were additionally calculated for the portion of the gait cycle corresponding to
17 hip flexion, between hip neutral and maximum hip flexion angle. Scores for the RMDQ
18 were calculated as the sum of ticked statements. Physical activity data was used to
19 ensure matching of subjects in LBP and NBP groups.

20 21 2.4 Statistical Analysis

A mixed-model analysis of variance (ANOVA) was used to compare within gait cycle peaks and degrees of hip flexion between LBP and NBP group data using SPSS (version 24, IBM Statistics, U.S.). *Post hoc* analyses for between groups comparisons were performed using independent samples t-tests where significant interactions were determined. Passive hip extensor moments and total hip moments, power and work done were compared at 25% increments of hip flexion, between neutral and maximum hip flexion. Total hip moments were additionally compared at FL, Ext1 and Ext2, and total power and work done were compared at H1, H2 and H3. Additional comparisons included age, gender, body mass, physical activity type and duration, walking speed and step length, maximum hip flexion angle, hip extension angle and knee angle, assessed using independent t-tests. CMCs and CVs were used to assess intra-subject gait cycle characteristics. RMSE and the adjusted r-squared were used to assess goodness of fit of the predictive equation coefficients for calculating passive moment contributions. Normality of data was assessed using Shapiro-Wilk tests, and the alpha level for all tests was set at 0.05.

3 Results

Subjects in the LBP group reported occurrence of back pain for 6.97 (SD 5.98) years with a range of 6 weeks to 30 years. RMDQ scores were 3.92 (SD 3.1) and VAS scores were 5.06 (SD 2.27). Independent samples t-tests demonstrated no significant differences between LBP and NBP groups for age, gender, body mass (within gender), walking speed or step length ($P > 0.05$). There were no significant differences ($P > 0.05$) in any of the

physical activity parameters calculated, which included walking (13.6 (SD 18.3) hours per week LBP, 9.9 (SD 12.9) hours NBP), moderate intensity exercise (6.9 (SD 11.0) hours per week LBP, 4.1 (SD 5.6) hours NBP) and vigorous exercise (6.7 (SD 8.7) hours per week LBP, 5.1 (SD 3.4) hours NBP).

The CMC means were 0.955 (SD 0.037) for LBP and 0.937 (SD 0.051) for NBP, respectively. The CV means were 21.4% (SD 11.6) and 26.3% (SD 13.9), for LBP and NBP, respectively. The RMSE was 0.016 (SD 0.008) Nm/(kg*m) for LBP and 0.017 (SD 0.07) Nm/(kg*m) for NBP. The adjusted r-squared values were 0.889 (SD 0.081) for LBP and 0.896 (SD 0.075) for NBP. There were no significant differences in hip or knee angles between groups ($P > 0.05$).

There were no statistically significant interaction effects in the total hip flexor (FL) or extensor moment peaks (Ext1, Ext2) ($F(1.571, 155) = 3.019, P = 0.065$, table 1) or their timings ($F(1.599, 158) = 0.368, P = 0.645$) (figure 2). There was no main effect of group ($F(1, 99) = 0.208, P = 0.649$). During the hip flexion component of the gait cycle, there was no statistically significant interaction effect in total hip moments ($F(2.064, 204) = 0.823, P = 0.444$) (table 2). There was no main effect of group ($F(1, 99) = 0.012, P = 0.913$).

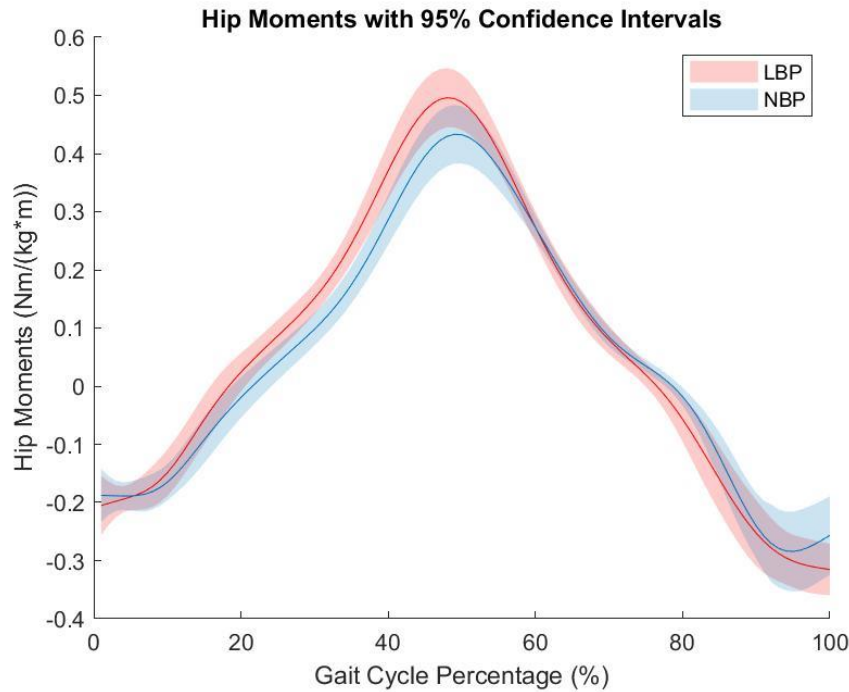


Figure 2. Mean total hip moments with 95% confidence intervals in LBP and NBP groups during gait cycle.

For passive hip moments there was a significant interaction effect of group and percentage of hip flexion ($F(1.751, 173) = 9.316, P = 0.000$) and a statistically significant main effect of group ($F(1, 99) = 6.597, P = 0.012$). Passive hip moments decreased throughout hip flexion, demonstrating increased extensor moments (figure 3), with no significant differences between groups at hip neutral or 25% ($P > 0.05$). From 50-100% of hip flexion, passive moments were significantly more negative in LBP subjects ($P < 0.05$), demonstrating greater extensor moments. At maximum hip flexion, passive hip extensor moments were 46.6% and 39.4% of total hip moments, for LBP and NBP, respectively. Passive hip extensor and total hip moment mean, standard deviation and results of the independent t-tests are shown in table 2.

Table 1. Variable data for complete gait cycle. NS (not significant) indicates ANOVA $P > 0.05$ with *post hoc* tests not indicated. Statistically significant results are denoted with an asterisk(*).

Moments (Nm/(kg*m))			Confidence Interval		ANOVA	Pairwise
		Mean (SD)	Lower	Upper	P-value	P-value
FL	LBP	0.557 (0.169)	0.51	0.604	0.065	NS
	NBP	0.482 (0.174)	0.435	0.529		
Ext1	LBP	-0.317 (0.122)	-0.351	-0.282		NS
	NBP	-0.278 (0.119)	-0.31	-0.245		
Ext2	LBP	-0.434 (0.153)	-0.477	-0.391		NS
	NBP	-0.421 (0.221)	-0.481	-0.361		
Power (W/(kg*m))						
H1	LBP	0.280 (1.96)	0.225	0.335	0.027*	0.99
	NBP	0.281 (0.159)	0.237	0.324		
H2	LBP	-0.520 (0.231)	-0.585	-0.456		0.041*
	NBP	-0.429 (0.212)	-0.487	-0.372		
H3	LBP	0.844 (0.298)	0.761	0.927		0.045*
	NBP	0.736 (0.235)	0.672	0.8		
Work Done (J/(kg*m))						
H1	LBP	0.028 (0.028)	0.02	0.036	0.010*	0.329
	NBP	0.034 (0.027)	0.026	0.041		
H2	LBP	-0.111 (0.059)	-0.128	-0.094		0.034*
	NBP	-0.087 (0.054)	-0.101	-0.072		
H3	LBP	0.115 (0.033)	0.106	0.124		0.017*
	NBP	0.101 (0.026)	0.094	0.108		

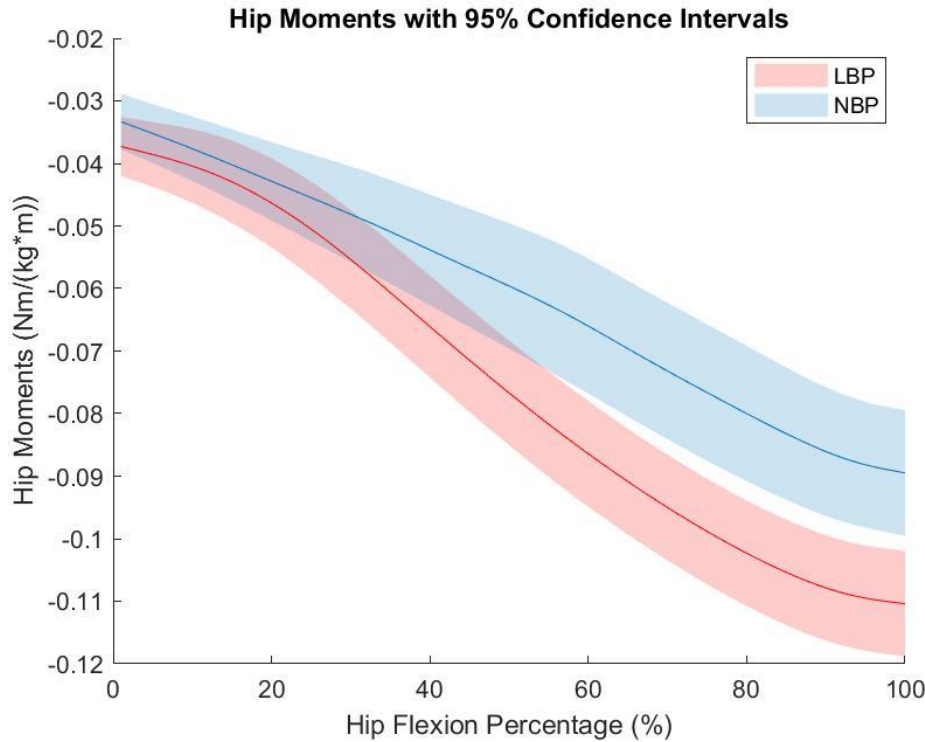


Figure 3. Mean passive hip extensor moments with 95% confidence intervals in LBP and NBP groups during hip flexion from neutral to maximum hip flexion

Table 2. Passive hip extensor and total hip moments (Nm/(kg*m)) during hip flexion, from neutral to full hip flexion. NS (not significant) indicates ANOVA $P > 0.05$ with *post hoc* tests not indicated. *P*-values included represent results of *post hoc* independent t-tests. Statistically significant results are denoted with an asterisk(*).

			Moments (Nm/(kg*m))			Pairwise <i>P</i> -value
			Mean (S.D.)	Lower	Upper	
0	Passive	LBP	-0.037 (0.017)	-0.042	-0.033	0.225
		NBP	-0.033 (0.016)	-0.038	-0.029	
	Total	LBP	0.228 (0.083)	0.204	0.251	NS
		NBP	0.194 (0.080)	0.172	0.216	
25	Passive	LBP	-0.050 (0.027)	-0.058	-0.043	0.345
		NBP	-0.046 (0.026)	-0.052	-0.039	
	Total	LBP	0.075 (0.069)	0.056	0.094	NS
		NBP	0.074 (0.073)	0.054	0.094	
50	Passive	LBP	-0.077 (0.030)	-0.085	-0.068	0.012*
		NBP	-0.060 (0.037)	-0.07	-0.05	
	Total	LBP	0.011 (0.043)	-0.002	0.023	NS
		NBP	0.020 (0.044)	0.008	0.032	
75	Passive	LBP	-0.099 (0.03)	-0.107	-0.09	0.002*
		NBP	-0.077 (0.040)	-0.088	-0.066	
	Total	LBP	-0.141 (0.168)	-0.188	-0.094	NS
		NBP	-0.116 (0.167)	-0.162	-0.071	
100	Passive	LBP	-0.110 (0.030)	-0.118	-0.102	0.002*
		NBP	-0.089 (0.037)	-0.099	-0.079	
	Total	LBP	-0.236 (0.192)	-0.29	-0.183	NS
		NBP	-0.226 (0.257)	-0.296	-0.157	

There was a statistically significant interaction of group and peak of hip power ($F(1.353, 134) = 4.366$, $P = 0.027$) but no main effect of group ($F(1, 99) = 0.06$, $P = 0.807$). There were no significant interactions of angle at peak ($F(1.526, 151) = 0.863$, $P = 0.398$). The first peak of power generation (H1) occurred at 8.2 (SD 4.5) %, and 10.0 (SD 4.4) % of gait cycle, for LBP and NBP, respectively, with no significant differences in timing or power ($P > 0.05$, table 1, figure 4). The hips then absorbed power, with a peak of power absorption (H2) during mid-stance. The peak of power absorption at H2 was significantly greater in LBP than NBP (-0.520 (SD 0.231) W/(kg*m), -0.429 (SD 0.212) W/(kg*m), for

LBP and NBP, respectively, $P = 0.041$), with no difference in timings between groups ($P > 0.05$). Power generation peaked at toe-off and the initiation of swing phase, at approximately 60% GC (H3). H3 was significantly greater in LBP than NBP (0.844 (SD 0.298) W/(kg*m), 0.736 (SD 0.235) W/(kg*m) for LBP and NBP, respectively, $P = 0.045$), with no difference in timings between groups ($F(1.526,151) = 0.863$, $P = 0.398$).

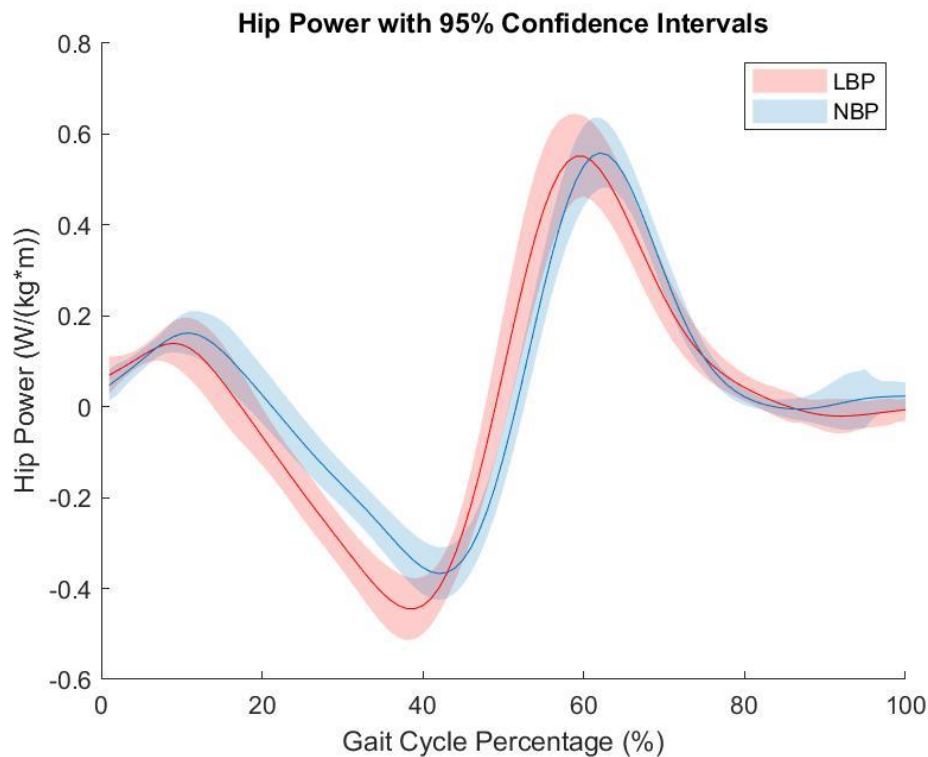
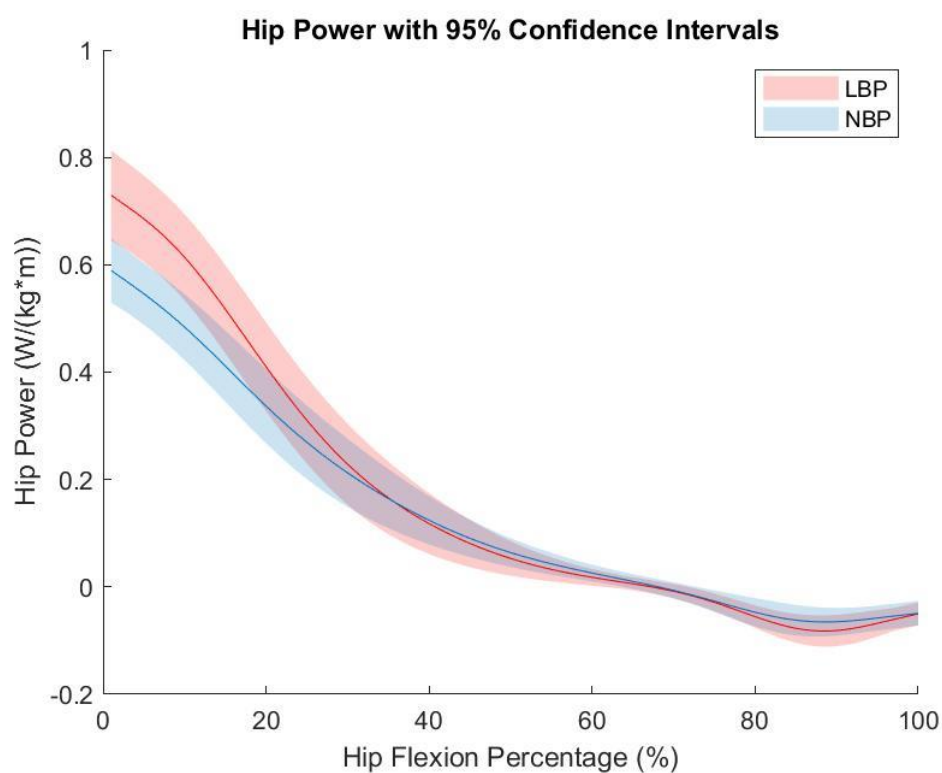


Figure 4. Mean total hip power with 95% confidence intervals in LBP and NBP groups during gait cycle.

For power during the hip flexion component of gait (figure 5), ANOVA results demonstrated a significant interaction of group and time ($F(2.264,224) = 3.228$, $P = 0.036$) but no main effect of group ($F(1,99) = 2.538$, $P = 0.114$). *Post hoc* analyses determined that total hip power was significantly greater ($P = 0.012$) in LBP subjects when the hip was in neutral (LBP = 0.717 (0.300) W/(kg*m), NBP = 0.583 (0.22) W/(kg*m). There were

1 no statistically significant differences between groups at any other percentage of hip
2 flexion assessed ($P > 0.05$). Mean, standard deviation and results of the independent t-
3 tests for hip flexion are shown in table 3.

4



5

6 Figure 5. Mean hip power with 95% confidence intervals in LBP and NBP groups during
7 hip flexion from neutral to maximum hip angle.

8

9 Table 3. Total hip power (W/(kg*m)) during hip flexion, from neutral to full hip flexion.
10 Statistically significant results are denoted with an asterisk(*).

		Mean (S.D.)	Power (W/(kg*m)) Confidence Interval		ANOVA <i>P</i> -value	Pairwise <i>P</i> -value
			Lower	Upper		
Hip Flexion Angle (%)	0	LBP	0.717 (0.300)	0.633	0.036*	0.012*
		NBP	0.583 (0.220)	0.523		
	25	LBP	0.309 (0.287)	0.228		0.455
		NBP	0.269 (0.252)	0.2		
	50	LBP	0.053 (0.115)	0.021		0.608
		NBP	0.064 (0.098)	0.037		
	75	LBP	-0.027 (0.059)	-0.043		0.928
		NBP	-0.026 (0.074)	-0.046		
	100	LBP	-0.054 (0.076)	-0.075		0.868
		NBP	-0.051 (0.085)	-0.074		

There was a significant interaction of group and peak for work done during gait ($F(1.263, 125) = 6.114$, $P = 0.010$), but no main effect of group ($F(1, 99) = 2.538$, $P = 0.114$). Negative mechanical work done during the H2 power absorption curve was greater in LBP than NBP (-0.111 (SD 0.059) J/(kg*m), -0.087 (SD 0.054) J/(kg*m), for LBP and NBP, respectively, $P = 0.034$). Positive mechanical work was greater in LBP than NBP during the H3 power generation curve (0.115 (SD 0.033) J/(kg*m), 0.101 (SD 0.026) J/(kg*m), for LBP and NBP, respectively, $P = 0.017$). During H1 and the hip flexion component of the gait cycle there were no statistically significant differences in work done between groups ($P > 0.05$). Total work done mean, standard deviation and results of the independent t-tests for hip flexion are shown in table 4.

Table 4. Total hip mechanical work done (J/(kg*m)) during hip flexion, from neutral to full hip flexion in 25% intervals and for overall hip flexion

		Work Done (J/(kg*m))			<i>P-value</i>
		Mean (S.D.)	Lower	Upper	
Hip Flexion Angle (%)	25	LBP	0.038 (0.017)	0.033	0.096
		NBP	0.031 (0.015)	0.027	
	50	LBP	0.009 (0.012)	0.006	
		NBP	0.010 (0.011)	0.007	
	75	LBP	0.001 (0.003)	0	
		NBP	0.001 (0.004)	0	
	100	LBP	-0.006 (0.007)	-0.007	
		NBP	-0.005 (0.007)	-0.006	
	Total Flexion	LBP	0.043 (0.028)	0.036	
		NBP	0.038 (0.028)	0.03	

4 Discussion

The present study found no statistically significant differences in hip and knee angles between groups. Values of passive hip extensor moments, and total hip moments, power and work done for NBP subjects were similar to those reported elsewhere [24]. Other investigators have reported alterations in hip and knee angles during walking in LBP subjects compared with healthy controls [25-27]. The most likely cause of increased passive moments in LBP is an increased passive resistance of the hip extensor and knee flexor muscles, such as the bi-articular hamstrings muscles. However, a previous study using the same subjects demonstrated no significant differences in passive moments between groups. Because passive hip moments increase with hip angle in an exponential manner, it is likely that small increases in hip angle have a considerable influence on

1 passive moments during walking, as observed in the present study. It is also plausible
2 that individual alterations and interactions of joint angles and passive moments may be
3 masked by averaging individual and group leg data. Alternatively, the difference in
4 passive moments equates to a maximum of 3 Nm in absolute rather than normalised
5 figures, which may be statistically significant but of little, if any, clinical relevance. Altered
6 passive moments occurred without differences in total moments.

7
8 Power absorption and work done at H2 were significantly greater in LBP than NBP.
9 The H2 power curve occurs during the stance phase of gait, as the hip flexors
10 eccentrically contract to stabilise the pelvis and support the trunk, preventing backwards
11 movement of the trunk relative to the pelvis. Power generation and work done at H3 were
12 also greater in LBP than NBP, and occurs during the late stance (pre-swing) phase of
13 gait, as the hip flexors contract concentrically to pull the leg forward and facilitate toe-off.
14 This coincided with greater power generation and work in LBP than NBP as the hip moved
15 through neutral.

16 The results of the present study demonstrate increased passive extensor moments
17 in LBP, with no differences in total moments. Whether the passive moments contribute
18 to total moments, thereby reducing the need for active contraction and promoting
19 efficiency, or if values of total moments should be altered due to the influence of greater
20 passive moments, is not currently understood [24]. Alternatively, altered joint angles, and
21 therefore passive moments, later in the swing phase may follow the increased total power
22 generation from the hip flexors during early swing, directly following toe-off. In this case,
23 increased passive moments may be due to increased joint angles, resulting from greater

1 momentum of the hip where active power generation is greater in LBP with no differences
2 in passive moments following toe-off. Greater momentum of the swinging limb could
3 cause greater hip flexion angles, contributing to greater passive resistance, and therefore
4 higher passive moments. The lack of difference in total moments may reflect reduced
5 active components late in swing, or be due to total moments being miscalculated due to
6 lack of account of passive influences in inverse dynamic modelling. Despite our lack of
7 understanding regarding the mechanisms and interactions of the passive and active
8 components, it is clear the passive component is considerable, and should not be
9 overlooked in future models of dynamic movements.

10 LBP subjects in the present study reported RDMQ scores that were low (3.92 (SD
11 3.1)) compared with other studies, VAS scores were considered moderate (5.06 (SD
12 2.27)), and similar to those reported elsewhere [28-35]. Although self-reported pain is a
13 subjective measure, subjects in the studies where VAS and RMDQ scores were higher
14 were recruited from physical therapy and rehabilitation clinics [28, 29, 31]. Overall, the
15 findings in the present study remain appropriate to the population assessed. Whilst LBP
16 severity and disability may be lower than in subjects attending rehabilitation clinics, there
17 were clear adaptations in gait to modify passive and active joint biomechanics compared
18 with healthy controls.

19 The present study indicates that passive moments may be influenced without
20 affecting total joint moments. Specifically, it is plausible that LBP may affect the structural
21 components of muscle tissue, or alternatively results in subtle interactions of joint angle
22 and moments during movement, sufficient to influence passive components without
23 affecting total moment. There may be further interactions between LBP, passive

1 moments, total power and total work done, which affect accelerations and energy
2 efficiency for movement. From a clinical perspective, an assessment of passive
3 biomechanics will be useful to help understand which tissue types are influenced by LBP,
4 on a case-by-case basis. Findings suggestive of structural adaptations will necessitate a
5 muscle-targeted therapy, such as stretching or strengthening exercise. Conversely, if
6 joint kinematics are altered, this indicates the active component of the muscular system
7 may have been affected, necessitating a therapy that targets neuromuscular
8 improvements, such as muscle activation timing and magnitude, and the specific fibres
9 recruited during movement.

10 A limitation of the current study was the measurement of hip passive moments
11 during hip flexion only. In previous studies [24, 36, 37], researchers averaged hip
12 moments during flexion and extension, and reported only a small underestimation of
13 moments during flexion, and overestimation during extension. However, in the present
14 study it was considered more appropriate to determine accurate values of passive hip
15 extensor moments, for integration into the hip flexion model during walking, rather than
16 over- or underestimating moments. Future studies would benefit from direct
17 measurement of both passive hip extensor and hip flexor moments to model the complete
18 gait cycle accurately in the sagittal plane.

19 Hip extensor moments were 46.6% and 39.4% of total hip moments for LBP and
20 NBP, respectively, at maximum hip flexion angle. This finding demonstrates that passive
21 structures may interact with active components, have a considerable influence on total
22 hip moments during walking [24], and should be integrated into future biomechanical
23 models. Whether it is appropriate to add passive moments to total moments, or to

subtract them, or if there is a more complex interaction between passive and active components to total moments is not currently understood [24] and therefore warrants further investigation.

Conclusions

The present data demonstrates that subjects with LBP have altered passive hip extensor moments and total power and work done during walking, compared with healthy controls. Although it is not possible to extrapolate cause and effect relationships, rehabilitation programmes for LBP patients should differentiate between the active, neuromuscular components of movement, and the passive components. Biomechanical models should include individual measurements of passive joint moments. The approach used in the present study may be a useful measurement model for clinicians assessing low back pain.

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